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THE EFFECTS OF NEODYMIUM-YTTRIUM-ALUMINUM-GARNET LASER IRRADIATION ON DYED AND UNDYED INSTRUMENTED ROOT CANAL WALL DENTIN IN HUMAN TEETH

by

© DOUGLAS N. DEDERICH

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE OF MASTER OF SCIENCE

DEPARTMENT OF ELECTRICAL ENGINEERING

EDMONTON, ALBERTA

SPRING, 1985

THE UNIVERSITY OF ALBERTA

RELEASE FORM

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THE EFFECTS OF NEODYMIUM-YTTRIUM-ALUMINUM-GARNET LASER IRRADIATION OF DYED AND UNDYED INSTRUMENTED ROOT CANAL WALL DENTIN IN HUMAN TEETH

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Date.....7/31/85.....

DEDICATION

To my parents, without whose love and support none of this could have been accomplished.

ABSTRACT

While the effects of lasers on tooth structure has been studied for a number of years, very little research exists concerning lasers with respect to their use in endodontics. As a prerequisite to developing an acceptable and beneficial technique utilizing lasers in endodontic therapy, it is necessary to develop an understanding of the effects of laser irradiation on the site which would be the primary target area in vivo, i.e. the instrumented canal wall dentin.

Twenty-two intact single rooted human teeth were split labio-lingually (longitudinally) from the incisal edge to the apex utilizing a thin carborundum disc and a slow speed handpiece. The root canal concavities of all specimens were then gently debrided using a round bur. At a series of evenly spaced locations along the length of the debrided canal wall, the concavity was flattened by an inverted cone bur to form small circular areas which were utilized as target locations. Eight specimens were stored in a thymol and water solution, subsequently rinsed in a 5.25% solution of sodium hypochlorite and water for one minute before being blotted dry and then irradiated by the laser. Thirty-five specimens were dyed with black India ink and allowed to air dry before irradiation. SEM analysis was performed.

The effects of the Nd-YAG laser on undyed and hydrated tooth specimens varied in such a way as to render statistical analysis impractical. There appeared to be a trend for the sclerotic and translucent areas of the dentin to pass the laser energy without noticeable thermal changes, and for the opaque dentin to absorb the energy to some degree and undergo thermal effects. These effects varied from no effects to disruption of the smear layer to actual melting and recrystallization of the dentin. The recrystallized canal wall dentin appeared non-porous and continuous in nature.

There was no statistically significant relationship between either the length of pulse duration or power level and the amount of thermal cracking or amount of fusion coverage in dyed and dehydrated tooth specimens when they were irradiated on their root canal wall with a Nd-YAG laser. However, there appeared to be a trend for shorter pulse durations to result more often in minimal thermal cracking and for median power levels to result more often in moderate to severe thermal cracking of the canal wall dentin.

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INTRODUCTION

1. The Problem

The dental pulp is the tissue from which most of the hard structure in teeth develops and which eventually resides at the center of the tooth surrounded by these same hard tissues. Its healthy composition is characterized by the presence of small blood vessels, nerves and connective tissue. Unfortuneately, the pulp possesses only limited ability to heal itself when injured or irritated. Dental caries, trauma and operative dentistry procedures are among the many factors which can contribute to pulpal pathology. When pathologic changes become severe and irreversible, the pulp must be removed and the space left by its absence, called the root canal space, must be obliterated. The act of removing the pulp and filling its space is referred to as the instrumentation and obturation of the root canal system. The consequences of failing to treat or to improperly treat root canals containing irreversibly damaged pulp tissue are continued pulpal degeneration and the eventual development of periapical pathology. Pain and infection are frequently associated with this. If root canal therapy is not implemented, then the extraction of the tooth will be an inevitable necessity at some time in the future. If root canal therapy is performed it is likely that pulpal pathology will not be among the reasons to remove the tooth should that eventually be required.

Proper instrumentation and obturation of the root canal system are the keys to successful endodontic therapy. Because of the biologic and morphologic complexity of the human root canal system as well as the environment which it presents and in which it exists, neither of these procedures is easily performed. Removing all organic tissues from the root canal space during instrumentation and attaining a fluid-tight seal with currently available obturation methods is easier said than done and has been demonstrated in several published articles[1-11]. Failure to completely debride the canal leaves necrotic tissues behind which may serve as substrates for bacterial growth. The failure to create a fluid-tight seal allows fluid to leak into the root canal. Potentially, it also provides a path for bacterial migration to the canal space and for microorganisms, their byproducts and other irritants to exit into the periapical tissues[12]. Since the pulpless tooth does not possess a viable circulation, the body's blood-borne defense mechanisms cannot effectively remove microorganisms and remnants of necrotic tissue from the root canal. It has been shown with <u>in vivo</u> animal research that a hollow space containing necrotic tissues and microorganisms can lead to pathologic changes in the surrounding tissue[13,14]. It therefore would appear both desirable and prudent, from the perspective of proper biologic rationale and past clinical experience, to perform the most effective instrumentation and obturation possible to insure the highest rate of success.

There are several things which can complicate and increase the difficulty of performing endodontic therapy. One of these is the decrease in the size of the pulp seen as a result of sclerosis. Other factors include the frequent necessity to treat posterior multicanal teeth where constricted and curved canals occur and access is cumbersome. As a response to these and other difficulties seen in many of the teeth requiring endodontic therapy, many dentists have chosen to utilize inferior but more convienient techniques. These techniques are not supported by any biologic rationale based on the current state of knowledge, do not result in the most effective instrumentation and obturation possible, are not taught in North American dental schools, and which can lead to serious pathologic sequela[15-19]. Typically, they require little precision or finesse and utilize tissue fixation chemicals. It has been shown that when such chemicals are used on vital pulpal tissues the remaining tissues seldom remain in or heal to a normal histologic state[20,21]. Accordingly, the substitution of a pulpal tissue fixation technique for traditional root canal therapy is considered to be biologically unacceptable.

Several studies have indicated that traditional endodontic therapy techniques yield a success rate ranging between 82 and more than 90 percent[22-25]. Thus, there seems to be room for significant improvement even over traditional techniques taught in

accredited dental schools. It should be noted that the success rates mentioned here are based on studies using relatively short recall intervals. It is possible that over a longer period of time (i.e. years) these rates are actually lower than indicated.

The difficulty observed in attaining complete canal instrumentation and obturation, and the propensity of some dentists to avoid traditional therapy techniques because of their complexity accentuates the need for continued develpoment of instruments, materials and techniques which are both more effective and less complex with respect to their application.

2. The Laser

Since the advent of the laser in the early 1960's, many applications for this technology have been derived and implemented. Although these include several accepted treatment procedures in the field of medicine, there are no presently accepted and widely practiced procedures utilizing laser technology in the field of dentistry. This is due to a number of reasons, not the least of which is the historical lack of comprehensive effort to develop such techniques and procedures. Some research does exist, however. This research tends to concern itself largely with the effects of the laser on both hard and soft tissues, and does not go into any great detail describing possible clinical techniques or applications. This reluctance to recommend the use of the laser on humans in the clinical setting is wisely based on

the fact that much more research is required before it can be considered for use. More in vitro and in vivo research is required in the following areas, to name but a few:

1. Microbiology: the effect of laser irradiation on the oral flora;
2. Histology: the effect of laser irradiation on the oral tissue;
3. Clinical Applications: whether the effects of the laser on the oral tissues can be controlled to provide clinically beneficial results;
4. Engineering: whether the appropriate laser output can be delivered to the target tissue in an adequately precise and safe manner;
5. Economics: whether the cost of laser equipment is justifiable when compared to the benefits of its use.

Attempts have been made to seal the apical root canal foramen externally at the root apex with the CO_2 laser and internally from within the root canal with the CO_2 and Nd-YAG lasers[26,27]. These attempts were unsuccessful in that charring and "cap" formation without fusion to the underlying dentin occurred and the foramina were not sealed. To date, with the exception of these preliminary and limited investigations of laser applications in endodontics, laser research in root canal therapy has been conspicuous by its absence.

A number of articles have been published which report dental hard tissue composition structural changes, soft tissue effects, and caries inhibition potentials[28-32]. Of particular interest is the report by Kantola[31] of dentin recrystallization after lasing to closely resemble normal enamel hydroxyapatite crystalline structure. Melcer et al.[33] has reported similar dentin crystalline structural changes as well as microhardness increases, and sterilization of dentin. This would indicate that appropriate laser application may induce crystalline changes in dentin which could be beneficial in endodontic therapy by effecting a decrease in dentin permeability. As previously mentioned, the amount of published research relating to the enhancement of canal sealability by the utilization of lasers is minimal. Those studies that do exist do not look at canal wall dentin in particular[30,31]. Canal wall dentin has been shown to vary both morphologically and histologically both within and between canals[34]. Accordingly, it is conceivable that the canal wall dentin may vary in its response to laser irradiation depending on the location of the target site. Given the potential of laser induced changes in dentin to decrease permeability, it follows that there exists a need to quantify the exposure parameters (i.e. peak power and pulse duration) required to consistently create this desired effect on canal wall dentin in particular.

This study examines the effects of Nd-YAG laser irradiation on both dyed and undyed sections of instrumented canal wall

dentin. Observation is performed utilizing scanning electron microscopy. Furthermore, nonparametric statistical analysis is made to aid in the identification of trends relating the desired effects (i.e. changes from porous or smear layer appearance to a "glazed" and nonporous appearance) to their corresponding laser exposure parameters.

LITERATURE REVIEW

The literature dealing with the topic of lasers in endodontics is scarce. However, research on lasers in other areas of dentistry has significant import with respect to laser use in endodontics. Furthermore, the economic feasability of the routine utilization of lasers in dentistry will likely depend on the degree to which it becomes useful and practical in other areas of dentistry as well. For these reasons, this wider area of research will be included in the following literature review.

ENAMEL

The effect of laser energy on enamel is the subject of much of the research that does exist concerning lasers in dentistry. A prominent reason for the interest that has been shown in this area is the potential of the laser to become an important tool in the practice of preventive dentistry. Although other aspects have been studied, such as the utilization of the laser as a dental drill, the majority of the research on enamel has been oriented to the field of preventive dentistry.

In 1964, Stern and Sognnaes[35] preliminarily investigated the effects of laser beam impact on dental hard tissues. Glass like fusion of enamel and charring of dentin were reported.

Efforts to fuse powdered substances with enamel were unsuccessful. The laser utilized was not specified.

In early preliminary studies performed in 1965, observations were made which indicated that influences on the enamel surface by exposure to laser beam irradiation may alter the relative resistance of enamel to dental caries as reflected in changes in its permeability, solubility or both[36-39]. In other words, the effect of a laser beam on the enamel surface might be either to render it less permeable to chemical agents by a physical fusion of the microstructure of the enamel surface or to render the quality of the enamel surface more amorphous or glazed so it is less soluble by providing the exposure of a smaller crystal surface area to the oral environment.

In 1966, Stern et al.[40] attempted to delineate the effect of the laser on the in vitro permeability and solubility of dental enamel. The authors found, again, that laser irradiation imparts a degree of impermeability to the enamel surface that reduced subsurface demineralization. No significant differences in solubility were found, however, between laser-exposed and laser-unexposed surfaces. The observed resistance of laser-exposed enamel to subsurface demineralization was attributed to an alteration in permeability rather than solubility. The laser utilized was not mentioned in this study.

In 1967, Gordon[41] attempted, unsuccessfully, to cut single surface cavity preparations in extracted teeth using a pulsed ruby laser. He did not attempt to determine the effects of the laser on the pulp.

Lobene et al.[42] in 1968 exposed extracted teeth to carbon dioxide laser radiation. He demonstrated disrupted enamel rods and incineration of odontoblastic processes in dentin. He attributed the observed changes to the local raise in temperature created by the laser. In addition, he noted that the X-ray diffraction pattern from lased enamel was almost identical to that corresponding to the unlased enamel. He accounts for what differences there were by suggesting the presence of α-calcium orthophosphate in the irradiated enamel, and not in the unirradiated enamel.

Scheinin et al.[43] produced craters in extracted teeth with a carbon dioxide laser in 1969. Energies great enough to penetrate the dentinoenamel junction were used. The authors described the charring of organic material, melted and vaporized enamel, and fissures all on a macroscopic basis. No further analysis was offered.

In 1972, Kantola[44] studied the changes in the calcium and phosphorous contents caused by intense carbon dioxide laser irradiation of enamel. He used both an X-ray microanalyzer and

polarized light microscopy. He found that the mineral distribution was not uniform; rather, a zonal pattern was observed in the mineral contents of the lased enamel reminiscent of a pressure wave moving in the direction of propagation of the laser beam.

Enamel exposed to laser irradiation was shown to be significantly more resistant to subsurface demineralization caused by environments conducive to dental caries by Stern et al.[45] in 1972. Also during that year, Stern et al.[46] showed that these effects could be achieved at much lower energy levels with a carbon dioxide laser when compared to a ruby laser.

In 1973, Kantola et al.[47] used X-ray diffraction to show that carbon dioxide laser induced effects produced recrystallization and growth of crystallite size in the hydroxyapatite crystals of dental enamel. The authors also concluded that, based on the X-ray diffraction pattern, there should be a small amount of α-calcium orthophosphate in lased enamel. α-calcium orthophosphate is the high-temperature form of the apatite phase of tricalcium phosphate.

Yamamoto et al.[48] in 1974 showed that low energy Nd-YAG laser irradiation imparted a degree of alteration to the enamel surface that subsequently reduced subsurface demineralization. Scanning electron microscopic observations of the laser irradiated enamel supported microradiographic findings and indicated that the

laser irradiated enamel surface was more resistant to decalcification that the nonirradiated control enamel. In vitro demineralization by lactic acid was accomplished by dextran producing strains of Streptococcus mutans PK-1 growing on the enamel surface.

In 1977 Goodman and Kaufman[49] studied the effects of an argon laser on the crystalline properties and rate of dissolution in acid of tooth enamel in the presence of sodium fluoride. The authors found that, upon exposure to laser radiation, enamel powder mixed with NaF (sodium fluoride) underwent an increase in crystalline size and/or perfection with a significant uptake of fluoride. Incisor teeth lased in the presence of NaF released significantly less calcium and phosphorous into sodium acetate (pH = 4.0) compared with unlased controls, suggesting a possible role for the laser in caries prevention. X-ray diffraction techniques were utilized to examine crystalline changes. Both this study and Stern et al.[40] suggest that lasing enamel in absence of NaF, although it does increase resistance to subsurface demineralization, does not alter solubility in acids. Goodman and Kaufman conclude here that lasing enamel powder in the presence of NaF will indeed decrease solubility in an acidic solution of sodium acetate.

A scanning electron microscopic comparison was made between human enamel surfaces treated with 50 percent phosphoric acid and

those subjected to an argon laser beam by Goodman and Gwinnett in 1977[50]. They found that preferential loss of rod structure and the creation of inter-rod and inter-crystal porosity were characteristic of acid etching. Pits, cracks, craters and blisters were found to be common topological features in lased enamel. Furthermore, the globular, frequently coalesced enamel rods and crystals differed from the spicular nature of these structures exposed by acid pretreatment. The lack of porosity and the presence of particulate surface debris following lasing did not allow resin penetration into enamel. In contrast, long resin tags were found following acid pretreatment. This observation would explain the findings of Winkler et al.[51] in which bond strengths for acid etched enamel were markedly superior to those for lased enamel.

DENTIN

The effect of laser radiation on dentin was studied simultaneously with the effects on enamel. Most of this research was limited to making various observations concerning the structural and chemical changes which occurred. To date, there has been relatively little research dealing directly with proposed clinical applications of the laser to dentin.

In a preliminary investigation in 1965, Kinersly et al.[52] used a pulsed ruby laser in an unsuccessful attempt to fuse dentin

and enamel powder. The authors also were able to burn holes in various other dental materials, including some metal alloys. Although attempts at welding these metals were not successful, it was concluded that, with improvement in laser technology, the laser holds potential for use in dentistry.

Also in 1965, Goldman et al.[53] noted that carious dentin was selectively destroyed by Q-switched ruby laser irradiation. It was pointed out, however, that it was still unclear if all of the carious dentin was removed or if it was only partially destroyed.

Peck et al.[54] in 1967 burned craters in both enamel and dentin with a pulsed ruby laser and, utilizing polarized light microscopy, described the effects as being in a zonal pattern. The authors also prognosticated on what potential uses the laser might have in dentistry as the technology was improved; however, they were unable to cite specifics concerning any particular clinical technique.

Vahl[55] bombarded both sound and carious dentin with ruby laser in 1968. X-ray diffraction analysis was done. Her results were limited to stating that melting of the dentin has indeed occurred, and that X-ray diffraction analyses showed the pattern of tricalcium phosphate.

Chimpanzee teeth were irradiated, in vivo, with a pulsed ruby laser in 1969 by Stern et al.[56] in order to explore the effect of such energy on the vital dental pulp. The authors utilized energy densities which were considered sufficient, by earlier studies, to decrease enamel susceptability to subsurface demineralization. The purpose of this study was to investigate whether irradiating enamel in such a way would create pulpal pathoses. They concluded that, within the limitations of the experiment, the pulpal changes observed after laser exposure were limited and reversible.

Adrian et al.[57] used a ruby laser to irradiate dogs' teeth to determine the energy density threshold level at which pulpal histologic or morphologic changes first appear. The authors determined this value to be 2,100 $\pm$ 200 joules/cm^2. It was also noted that at this level there was no measurable loss of enamel. It was not until much higher densities were used that a significant loss of tooth structure was noted. At these larger doses, however, pulpal necrosis was severe and consistent. It was concluded that, at that time, lasers were not suitable for the preparation of cavities.

In 1971, Weichman et al.[26,27] attempted to externally and internally seal the apical foramen of an extracted tooth utilizing a carbon dioxide laser. Both attempts were unsuccessful, and the authors therefore concluded that the use of the laser in

endodontics was not yet feasible.

In 1972 Kantola[30] analyzed dentin that had been exposed to carbon dioxide laser irradiation. Utilizing microradiography and electron probe microanalysis, it was determined that the calcium and phosphorous contents of the exposed dentin were clearly higher than those of normal dentin. Kantola attributes this to the burning off of the local organic material, leaving behind the inorganic phosphorous and calcium in higher concentrations per unit volume.

In another study, Kantola[31] analyzed the crystallographic changes in carbon dioxide lased dentin by means of X-ray diffraction. He found that both recrystallization and growth in crystal size occurred, and that the dentin thereby changed structurally in such a way that it came to closely resemble the crystalline structure of the hydroxyapatite of normal enamel.

In order to further investigate the pulpal response to laser irradiation, Adrian et al.[58] exposed eight teeth from two rhesus monkeys to various amounts of Nd-YAG laser energy. The pulp response was found to be much less severe than that in a study previously reported[57] for exposure to similar amounts of ruby laser radiation. This observation led Adrian to suggest that, if additional studies were to support this finding, a reevaluation of Nd-YAG laser procedures on vital teeth should be done since it may

be that pulpal response may not contraindicate its use.

In 1980, Yamamoto et al.[29] used an acousto-optically Q-switched Nd-YAG laser at low energy density in an attempt to inhibit caries formation in the enamel of extracted teeth. Fiber optic delivery was used, and a "laser absorption material" (which was not named) was coated on the target surfaces before lasing. The temperature rise of the pulp was also measured, utilizing a thermocouple. The investigators found that caries inhibition did indeed result from this exposure and that the temperature rise of the pulp was 20°C. They rationalized that although this temperature rise above body temperature (37°C to 57°C) will produce pulp destruction for exposures longer than ten seconds, it likely will pose no significant danger in this technique since the exposure time is less than one second and much of the energy is absorbed by the material placed on the enamel surface prior to lasing.

Brune[59] attempted to cut pin holes in extracted teeth in 1980 with a pulsed carbon dioxide laser. He found that holes created in such a manner were less than ideal due to cracks and a lack of geometrical symmetry. Furthermore, he found the Vickers hardness of the walls of these holes was similar to that of enamel.

SOFT TISSUE

Research concerning lasers and the oral soft tissues does not seem to be plentiful when compared to that which is available for the dental hard tissues, although this is not so concerning soft tissues of other parts of the body.

Yamamoto et al.[60] conducted a preliminary investigation in 1982 on the effects of ruby laser radiation on tongue and auricular soft tissue in rats. His observations were qualititive in nature with respect to the general histological changes seen. The investigators also concluded that clinical application of the laser in dentistry should still be considered premature.

WELDING

Interest in the use of lasers to weld dental prostheses has been considerable due to a number of reasons, one of which is time savings conceivable gained by the simplification of both intraoral preparation of abutment teeth and the actual joining (welding or soldering) process.

In 1970, Gordon and Smith[61] used a Nd-YAG laser to weld sections of various fixed and removable prostheses, <u>in vitro</u>. They summarized their observation by noting that a considerable amount of time was saved when laser welding was used when compared to the

conventional solder technique. They also found that inaccuracies in assembly caused by transfers from the master cast along with investment and heat distortions were avoided. A functional advantage was that the anatomic form of the interproximal embrasures is not obliterated when a laser is used. The authors also mentioned that esthetics can be maximized, especially in the anterior regions where so very important, because individual sculpturing of abutments and pontics can be accomplished without the restrictions enforced by solder joint design.

A metallurgical study of laser welded joints was performed in 1972 by Smith et al.[62]. Here, a neodymium-glass laser was used to produce weld joints between test pieces of four different casting alloys. The mechanical properties of the welds were determined and their microstructures were evaluated. The investigators found that laser welding is rapid and convenient, and it can provide sound welds in dental casting gold alloys. The strength of the weld was found to be comparable to the annealed strength of the actual alloys used. It was concluded that laser welding of the casting gold alloys used in dentistry is both practical and metallurgically feasible.

An attempt was made to fuse several currently available tooth colored restorative materials to enamel using a neodymium laser by Winkler et al.[63] in 1973. No evidence of fusion was observed microscopically and dye penetration at the enamel/

restoration interface in all cases indicated that lasing was ineffective in sealing off the junction between the restorative material and the tooth.

In 1975, Preston and Reisbick[64] compared conventional joining methods with laser fused precious, nonprecious and mixed welds and concluded that laser fusion is as good as, or better than, unions created by conventional dental joining methods.

Laser microwelding of orthodontic appliances was studied in a simulated intraoral environment by Colby and Bevis in 1978[65]. It was concluded that safe intraoral welding can be done with a pulsed neodymium laser and proper controls on a routine basis with no damage to dental hard tissues. However, improvements in laser technology and technique were implied to be necessary before widespread use of intraoral laser welding be condoned.

METHOD AND MATERIALS

Nd-YAG Laser: Undyed Specimens

1. Four intact single-rooted human teeth were split labio-lingually (longitudinally) from the incisal edge to the apex utilizing a thin carborundum disc and a slow speed handpiece. This exposed the walls of the root canal as the root canal was effectively divided between the two resulting halves of the tooth. Easy and controlled access to the total length of the canal wall was thereby facillitated.

2. The canal concavities in all of the eight specimens were then gently debrided along their entire length using a round bur. This removed any remaining necrotic tissue tags or organic debris.

3. At a series of evenly spaced locations along the length of the debrided canal wall, the concavity was flattened by an inverted cone bur to form small circular areas which were utilized as target locations (Figure 1 and Plate 1). These small, circular flattened areas are advantageous to create in this manner for three reasons:

a) flat areas require less manipulation of microscope controls to maintain proper focus as the specimen is moved about during SEM (scanning electron microscopic) observation,

b) the use of the inverted cone bur creates a dentinal smear layer on the surface of the dentin which simulates that which is left by instrumentation during traditional root canal therapy techniques, and

c) utilizing numerous small target preparations per specimen allows the efficient use of both the available acceptable tooth specimens and SEM time.

4. All specimens were stored in a thymol-water solution and subsequently rinsed in a 5.25% sodium hypochlorite solution for one minute. The utilization of 5.25% sodium hypochlorite solution mimics currently accepted techniques where irrigation of the canal is accomplished by flushing the canal space with such a solution.

5. The specimens were blotted dry, and each one-half tooth assigned its own label.

6. The eight specimens were then irradiated according to Table I. The Nd-YAG laser energy was delivered to target area via a 600 micron quartz optical fiber. The distal tip of the fiber was held approximately 2 mm away from the prepared

target surface by means of a hydraulic micrometer device, and the angle of incidence was perpendicular to this surface (Plate 2). The exact impact area was identified utilizing the aiming light. The Nd-YAG laser utilized was the Molectron 5000 Photocoagulator (Plate 3). All appropriate safety precautions were taken before commencing with this experiment.

7. After taking the necessary color photographs of the irradiated areas, the specimens were prepared for SEM evaluation.

Nd-YAG Laser: Dyed Specimens

The experimental protocol was the same for the dyed specimens as for the undyed specimens with the following exceptions:

1. Thirty-five one-half tooth specimens were utilized.
2. The specimens were stained with black India ink and allowed to dry before irradiation (Plate 4).
3. The specimens were irradiated according to Table II.

EVALUATION OF DATA

As previously mentioned, the goal of traditional endodontic therapy is to eradicate pulpal and periapical pathoses and prevent its recurrance, and this is usually facilitated by improving the canal seal. Evaluating the results of laser impacts on canal wall dentin by means of a scanning electron microscope is necessarily an indirect means because the observations that can be made are limited to the categorization of topographical features and do not involve direct measurements of sealability. The topographical features used to categorize the data in this study were:

a) amount of impact area covered by fused and recrystallized dentin, and

b) the amount of cracking due to thermal stresses in the target area.

Accordingly, the following procedure was used to evaluate the data:

1. SEM photographs were taken of every laser beam impact site. As a general rule, these photographs completely included the impact area and any surrounding laser-induced changes.

2. Opaque tape was placed over the identification labels on the photographs to avoid bias during evaluation based on evaluator knowledge of the power and pulse duration.

3. The photos were then randomly assigned numbers by a disinterested third party. A record of these assignments was made and kept secret from those who were to evaluate the photographs.

4. In order to control for dessication cracking due to specimen preparation necessary for SEM observation, two evaluators together examined all the photographs and marked all cracks that they both agreed were most likely due to dessication and not laser-induced thermal stresses.

5. Each evaluator previewed fifty randomly selected photographs to obtain a subjective impression of the range of the degree of thermal cracking and amount of the laser impact area actually fused.

6. Each evaluator then evaluated the photographs and categorized, subjectively, the data using the form shown in Figure 2. Although the order in which the photographs were evaluated was random, the evaluators observed them in the same random sequence.

7. After initial evaluation it was noted that disagreement existed between the two evaluators in 23 of the 98 photographs. These disagreements were reviewed in light of

the categories used and then were independently rejudged by both evaluators. This resulted in agreement in 18 of the 23 photographs. Discussion between the examiners on the last 5 photographs resulted in agreement on every case.

8. The data was then grouped and then ranked (with rank 0 corresponding to no or minimal thermal cracking or fusion coverage and rank 1 corresponding to moderate to severe thermal cracking and moderate to total fusion coverage) according to the charts shown in Figures 3,4,5 and 6 in preparation for the Friedman two-way analysis of variance by ranks. In theory, the total rank values in these figures are used to calculate a Chi-Square value, which, when compared to a table of critical values of Chi-Square for the given number of degrees of freedom, will give a significance level indicating the probability that the data occured purely by chance. For a more detailed description of this test, see reference 69 and the appendix.

RESULTS

1. Undyed Specimens

The results of Nd-YAG laser impact on the undyed canal wall dentin target areas are shown in Table III. All observations were made utilizing scanning electron microscopy. Only three exposures resulted in actual melting and recrystallization (fusion) of the dentin, and all three occurred in areas consisting largely of opaque dentin. Virtually no fusion occurred in areas of translucent dentin. All undyed specimens were well hydrated before irradiation. All fusion areas demonstrated minimal or no thermal cracking. Predictability of laser effects on undyed dentin was found to be poor. The data seemed to indicate a trend of fusion occurring only where the impact area consisted of opaque dentin. The small number of actual "successes" or fusion sites and the limited number of total exposures made further statistical analysis impractical, especially in view of the dominating effect of the varying optical properties of the target dentin on the absorption of the laser energy.

Laser induced disruption of the smear layer is demonstrated in Plate 5. An example of dentinal fusion is shown in Plates 6,7 and 8.

2. Dyed Specimens

The results of Nd-YAG laser impact on dyed canal wall dentin target areas are shown in Tables IV and V. All observations were made utilizing scanning electron microscopy.

The effects of Nd-YAG laser irradiation on a dyed specimen is demonstrated in Plates 9 and 10. Plate 9 shows an example of fusion with minimal cracking and Plate 10 shows an example of fusion with moderate to severe thermal cracking.

The Friedman's two way ANOVA by ranks was performed four times in total. The data for these tests was grouped first by power and then by time of pulse duration for both tests of thermal cracking and fusion coverage, resulting in four separate runs.

The statistical results from the Friedman's ANOVA of fusion coverage is shown in Tables VI and VII. The results from the Friedman's ANOVA of the amount of thermal cracking is shown in Table VIII and IX.

The analysis of the fusion coverage data failed to show a systematic pattern of the distribution of the mean ranks ($\alpha = .9821$ for data grouped by power, and $\alpha = .8266$ for data grouped by time).

The results of the analysis of the thermal cracking data was likewise not significant (assuming a desired confidence level of $\alpha = .05$ or less). However, there was a decrease in alpha levels

(α = .2145 for data grouped by power, and α = .1040 for data grouped by time) when compared with those resulting from the analysis of fusion coverage.

DISCUSSION

1. Rationale

The perceived need to "glaze over" and occlude the microtubules in canal wall dentin arises from the fact that these tubule spaces can harbor bacteria, their toxins and other irritants which conceivably can contribute to the failure of an endodontically treated tooth. Even the smear layer, which is a layer of dentin chips and canal debris 1 to 5 microns thick created by instrumentation of the canal, has been shown to be an imperfect barrier to the penetration of fluids into the tubule spaces over which it lies[66,67], albeit an improvement over the absence of a smear layer. This is opposite of what would be expected judging only from its smooth, continuous and nonporous appearance. Furthermore, the ability to melt and recrystallize dentin allows for the possibility of fusing apical plug materials intimately and continuously into the canal wall itself, thereby facilitating a fluid-tight apical seal as well.

The topographical features used to categorize the data in this study were chosen on the basis of their assumed effect on the actual leakage (i.e., permeability to fluids). In other words, it was assumed that 1) the greater the percentage of the impact area that was fused, and 2) the smaller the number and size of thermal

cracks all implied an increased likelyhood of reduced permeability to fluids. In this assumption lies the rationale for choosing the above two topographical features upon which to base the statistical test utilized in this study.

Due to the nature of the effects of laser irradiation on dentin, it was possible to quantify the observations with respect to the two topographical features only in general categories. That is, powerful interval measurements were not possible. This made it necessary to utilize nonparametric statistics rather than parameteric statistics, where the underlying statistical assumptions do not all apply to the data taken in this study[68]. Furthermore, it was possible to rank these categories in order of preference with respect to the assumed effect on permeability. Since the data was ordinal in nature and the observations were related by either pulse duration or power utilized, the appropriate statistical test was determined to be the Friedman two-way analysis of variance by ranks[69].

2. Sources of Error

Possible sources of error and inconsistency exist in the laser equipment and the manner in which it was utilized, the experimental protocol used and the morphological and histological variations of the target tissue.

Probably one of the largest sources of error was the variation of the focal spot size on the target surface caused by changes in the geometry and transmission ability of the distal end of the optical fiber. Being held only 2 millimeters away from the target surface, it was subjected to splattering by melted tooth structure caused by the impact of the laser energy on the target, to some of its own energy reflected back on itself and to the thermal effects associated with these two phenomena. This made it necessary to periodically cleave a new end on the fiber. However, for practical reasons, this was not done after every exposure. Conceiveably, then, variations in spot size not only probably occurred between the cleaving of new distal ends, but also between freshly cleaved ends since it is not likely two freshly cleaved ends will have exactly identical geometries. This variation in focal spot size (and thereby variation in incident energy density) may in part explain the almost random variation seen in the amount of fusion coverage (α = .9821 for data grouped by power, α = .8266 for data grouped by time; see Tables VI and VII).

The energy density is important insofar as it directly influences the temperature of the dentin surface. It can be approximated by dividing the total energy delivered by the surface area over which it was distributed.

$$\text{Energy Density} = \frac{\text{Total Energy Delivered}}{\text{Surface Area}} = \frac{(\text{Watts})(\text{Seconds})}{\text{cm}^2}$$

It is assumed here (as in the case of the dyed specimens) that the energy is actually localized to the surface. In the case of the undyed specimens, where the energy is distributed throughout a tissue volume due to the penetrating nature of the Nd-YAG laser, the total energy should be divided by the appropriate volume. The diameter of the focal spot (i.e. the spot size) was subjectively determined to average about 1 millimeter. Accordingly, the approximate impact surface area can then be calculated.

$$\begin{aligned} \text{Area} &= \pi r^2 \\ &= \pi(5 \times 10^{-2})^2 \\ &= \pi(2.5 \times 10^{-3}\ cm^2) \\ &= 7.85 \times 10^{-3}\ cm^2 \end{aligned}$$

The results of this study were not described in terms of the energy density because it was desired to analyze separately the effects of power and time of laser irradiation on the dentin. However, to facilitate comparisons with other studies which relate their results in terms of incident energy density, the corresponding energy densities for the powers and pulse durations used in this study can be found in Table X. Note that this table corresponds only to a spot size whose diameter is 1 millimeter and assumes all the energy was absorbed on the immediate surface of the dentin.

Although it was assumed that coating the target areas with India ink would cause the incident laser energy to be localized on

the surface, it is not known to what degree this occurs. Due to the tendency of the black ink to absorb energy and subsequently burn away, it is necessary to acknowledge the possibility of energy absorbtion (and therefore thermal effects and stresses) at deeper levels in the dentin after the dye is obliterated. It is also possible that variations in the thickness of the dried ink on the target surface could introduce additional inconsistencies by varying the time necessary to burn away the dye. The relevance of the tendency of the dye to burn away becomes obvious when one considers the situation where the dye is obliterated before the laser exposure is ceased, allowing the laser energy to interact directly with the dentin for a time without the intervention of the dye. It is conceiveable that, in this case, the penetrating nature of the Nd-YAG laser energy could not only increase the depth of thermal effects in the dentin, but also reach the surrounding periodontal ligament and other periradicular tissues.

Another possible source of inconsistency in both the undyed and dyed studies is the variation in the histology and morphology of the target dentin. Canal wall dentin has been shown to vary with respect to its tubular morphology both in the number (density) of tubules and their degree of sclerosis[34]. Preliminary observations have indicated, as did the undyed specimen results of this report, that sclerotic (i.e., translucent) dentin tends largely to pass or transmit Nd-YAG laser energy. On the other hand, dentin where the tubules are numerous and open (i.e., opaque

dentin) tends to absorb the incident energy to a greater degree and therefore undergoes observable thermal-induced changes earlier. These variations introduced enough error in the undyed specimen case as to render any statistical analysis of the data meaningless.

The degree to which the target specimen is hydrated may also be an important factor with respect to both the power and pulse duration required for dentinal fusion and for the degree of thermal cracking resulting from laser impact. Before melting and recrystallization of the inorganic component of dentin can occur, both the organic material and any water in the immediate area must first be vaporized. Accordingly, the presence of additional water may increase the overall energy requirements for fusion. Clinical experience has indicated that dehydrated and nonvital teeth seem to be more brittle than hydrated and vital teeth. If this increase in brittleness can be attributed to an increase in internal stresses brought about by dehydration, then it follows that more thermal cracking after laser irradiation may be seen in dehydrated specimens (as in the dyed specimen study) when compared to hydrated specimens (as in the undyed specimen study).

3. Trends

Due to the high alpha levels resulting from the analysis of fusion coverage ($\alpha = .9821$ for data grouped by power, and $\alpha = .8266$ for data grouped by time), it is likely that the given distribution of data was largely due to chance and correspondingly had only

minor influence from other factors such variation in spot size and thereby incident energy density due to fiber tip degradation and dehydration of the specimens before irradiation. The lower alpha levels seen with the analysis of thermal cracking ($\alpha = .2145$ for data grouped by power, and $\alpha = .1040$ for data grouped by time) may indicate that shorter pulse durations tend to result more often in less thermal cracking and that median power levels (10 to 26 Watts) tend to result more often in moderate to severe thermal cracking.

The trend for shorter pulse durations to result in minimal thermal cracking can be partly explained by the idea that time is required for significant heat flow (and therefore thermal effects) to occur, and the less time that is available the less the heat and thermal effects will penetrate the dentin.

The trend for moderate to severe cracking to occur more often at median power levels is more difficult to explain. It is concieveable that this may relate to the composition of the dentin and the manner in which it reacts to various temperatures. If the dentin is raised to temperatures above a certain threshold, the organic component (about 30%) will char, and cracks will form (especially in dyhydrated specimens). At yet a higher threshold, the local organic component will vaporize and the inorganic component will melt and flow, obliterating the cracks and leaving a smooth surface upon cooling. Heating the dentin to even a higher threshold vaporizes both the organic and inorganic components,

leaving the surface irregular, porous and perhaps also cracked. Laser power levels which could create this complete spectrum of temperatures were used in this study. If, indeed, the median power range tended to create temperatures which resulted in charring but not enough melted inorganic material to flow and obliterate the cracks, the above may in part explain the tendency for median power levels to result more often in moderate to severe thermal cracking.

CONCLUSIONS

The effects of the Nd-YAG laser on undyed and hydrated tooth specimens varied in such a way as to render statistical analysis impractical. There appeared to be a trend for the sclerotic and translucent areas of the dentin to transmit the laser energy without noticeable thermal changes, and for the opaque dentin to absorb the energy to some degree and undergo thermal effects. These effects varied from no effects to disruption of the smear layer to actual melting and recrystallization of the dentin. The recrystallized canal wall dentin appeared nonporous and continuous in nature.

There was no statistically significant relationship between either the length of pulse duration or power level and the amount of thermal cracking or amount of fusion coverage in dyed and dehydrated tooth specimens when they were irradiated on their root canal wall with a Nd-YAG laser. However, there appeared to be a trend for shorter pulse durations to result more often in minimal thermal cracking and for median power levels to result more often in moderate to severe thermal cracking.

RECOMMENDATIONS

In further similar studies, in order to render the statistical analysis more powerful, it is recommended that several (3 or 4) target areas be irradiated for each combination of power level and pulse duration rather than the single exposure utilized in this study. It is conceiveable that this may render more definitive and significant conclusions due to an increase in the information contained by the data.

Also, it is recommended that hydrated specimens be utilized in further studies on the effects of lasers on dentin since this is the situation which exists in vivo, and in view of the possibility that dehydration may severly increase the incidence of thermal cracking upon impact of laser radiation.

TABLES

TABLE I

UNDYED SPECIMENS

PULSE DURATIONS, SECONDS

Specimen	A	B	C	D	E	F	G	H
Power (Watts)	20	30	40	50	60	70	80	90
Target Location								
#1	.1	.1	.1	.1	.1	.1	.1	.1
#2	.2	.2	.2	.2	.2	.2	.2	.2
#3	.3	.3	.3	.3	.3	.3	.3	.3
#4	.4	.4	.4	.4	.4	.4	.4	.4
#5	.5	.5	.5	.5	.5	.5	.5	.5
#6	.6	.6	.6	.6	.6	.6	.6	.6
#7	.7	*	.7	.7	.7	*	*	.7
#8	*	*	.8	.8	.8	*	*	.8
#9	*	*	.9	*	*	*	*	.9

* Target Location was not used or does not exist on this particular specimen.

Note: Locations 1 through 9 proceed from the incisal-most portion of the canal to the apical-most portion.

TABLE II

DYED SPECIMENS

PULSE DURATIONS, SECONDS

Specimen	I1	I2	J1	J2	K1	K2	L1	L2
Power (Watts)	5	5	8	8	11	11	14	14
Target Location								
#1	.1	.7	.1	*	.1	.7	.1	.7
#2	.2	.8	.2	.7	.2	.8	.2	*
#3	.3	.9	.3	.8	.3	.9	.3	.8
#4	.4	1.0	.4	.9	.4	1.0	.4	.9
#5	.5	*	.5	1.0	.5	*	.5	1.0
#6	.6	*	*	*	.6	*	.6	*
#7	*	*	.6	*	*	*	*	*

* Target Location was not used or does not exist on this particular specimen.

Note: Locations 1 through 7 proceed from the incisal-most portion of the canal to the apical-most portion.

TABLE II (CONTINUED)

DYED SPECIMENS

PULSE DURATIONS, SECONDS

Specimen	M1	M2	N1	N2	O1	O2	P1	P2
Power (Watts)	17	17	20	20	23	23	26	26
Target Location								
#1	.1	.7	.1	.6	.1	.7	.1	.7
#2	.2	.8	.2	.7	.2	.8	.2	.8
#3	.3	.9	.3	.8	.3	.9	.3	.9
#4	.4	1.0	.4	.9	.4	1.0	.4	1.0
#5	.5	*	.5	1.0	.5	*	.5	*
#6	.6	*	*	*	.6	*	.6	*
#7	*	*	*	*	*	*	*	*

TABLE II (CONTINUED)

DYED SPECIMENS

PULSE DURATIONS, SECONDS

Specimen	Q	R	R	S	S
Power (Watts)	30	30	40	40	50
Target Location					
#1	.1	.8		*	
#2	.2	.9		.6	
#3	.3		.1		.1
#4	.4		.2		.2
#5	.5		.3		
#6	.6		.4		
#7	.7		.5		

TABLE III

LASER EFFECTS ON UNDYED CANAL WALL DENTIN

		PULSE DURATION, SECONDS								
		0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9
Target Location:		1	2	3	4	5	6	7	8	9
	20	0	0	0	0	0	0	d	*	*
	30	-	-	-	-	-	F	*	*	*
	40	0	0	0	0	d	d	d	d	F
POWER,	50	0	0	0	0	0	0	0	0	*
WATTS	60	0	0	0	0	0	0	0	0	*
	70	0	0	0	0	0	0	0	*	*
	80	d	d	0	0	0	0	*	*	*
	90	0	0	d	d	d	d	d	d	F

Key: 0 = no change

F = fusion, with melting, recrystallization and minimal to no thermal cracking

d = disrupted, dessicated smear layer

* = no such target location existed on this specimen

TABLE IV

RESULTS

LASER EFFECTS ON DYED CANAL WALL DENTIN

		PULSE DURATION, SECONDS									
		0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9	1.0
POWER, WATTS	5	1	1	0	1	1	1	1	1	1	1
	8	1	0	1	1	1	1	1	1	1	1
	11	0	1	1	1	1	1	1	1	1	1
	14	0	1	1	1	1	1	1	1	1	1
	17	1	1	1		1	0	1	1	1	1
	20	1	1	1	1	1	1	1	1	1	
	23	0	1	1	1	1		1	1	1	1
	26	1	1	1	1	1	1	1	1	1	1
	30	1	1	1	0	1	1	1	1	1	
	40	0	1	1	1	1	1				
	50	1	1								

ATTRIBUTE TESTED: FUSION COVERAGE

0 = None to Slight

1 = Moderate to Total

TABLE V

RESULTS

LASER EFFECTS ON DYED CANAL WALL DENTIN

		PULSE DURATION, SECONDS									
		0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9	1.0
	5	0	0	0	1	1	1	1	1	1	1
	8	0	0	1	1	1	0	1	1	1	0
	11	0	1	0	1	0	0	1	1	1	1
	14	1	1	1	1	1	1	1	1	0	1
POWER,	17	0	1	1		0	0	1	1	1	1
WATTS	20	0	1	1	1	1	1	1	1	1	
	23	1	1	1	0	0		1	1	1	1
	26	1	1	1	0	1	1	1	1	1	1
	30	1	1	1	1	1	1	1	1	1	1
	40	0	1	1	0	0	1				
	50	0	0								

ATTRIBUTE TESTED: THERMAL CRACKING

0 = None to Slight

1 = Moderate to Severe

TABLE VI

RESULTS

FRIEDMAN'S TWO WAY ANOVA BY RANKS

ATTRIBUTE TESTED: FUSION COVERAGE

DATA GROUPED BY PULSE DURATION

Variable (Power, Watts)	Mean Rank
1 - 10	2.50
11 - 15	3.00
16 - 20	3.00
21 - 26	3.25
27 - 50	3.25

Number of Cases: 10

Chi-Square: 1.5

Degrees of Freedom: 4

Significance: .8266

TABLE VII

RESULTS

FRIEDMAN'S TWO WAY ANOVA BY RANKS

ATTRIBUTE TESTED: FUSION COVERAGE

DATA GROUPED BY POWER

Variable (Pulse Duration, Seconds)	Mean Rank
0.1	4.00
0.2	5.00
0.3	5.00
0.4	6.00
0.5	6.00
0.6	5.00
0.7	6.00
0.8	6.00
0.9	6.00
1.0	6.00

Number of Cases: 5

Chi-Square: 2.4545

Degrees of Freedom: 9

Significance: .9821

TABLE VIII

FRIEDMAN'S TWO WAY ANOVA BY RANKS

ATTRIBUTE TESTED: THERMAL CRACKING

DATA GROUPED BY PULSE DURATION

Variable (Power, Watts)	Mean Rank
1 - 10	2.00
11 - 15	3.60
16 - 20	3.60
21 - 26	3.20
27 - 50	2.60

Number of Cases: 10

Chi-Square: 7.6800

Degrees of Freedom: 4

Significance: .1040

TABLE IX

FRIEDMAN'S TWO WAY ANOVA BY RANKS

ATTRIBUTE TESTED: THERMAL CRACKING

DATA GROUPED BY POWER

Variable (Pulse Duration, Seconds)	Mean Rank
0.1	2.60
0.2	5.40
0.3	5.70
0.4	5.30
0.5	3.60
0.6	4.80
0.7	7.50
0.8	7.50
0.9	6.50
1.0	6.10

Number of Cases: 5

Chi-Square: 11.9782

Degrees of Freedom: 9

Significance: .2145

TABLE X

ENERGY DENSITIES, J/cm^2

		PULSE DURATION, SECONDS				
		0.1	0.2	0.3	0.4	0.5
	5	63.7	127.4	191.1	254.8	318.5
	8	101.9	203.8	305.7	407.6	509.6
	11	140.1	280.3	420.4	560.5	700.6
	14	178.3	356.7	535.0	713.4	891.7
	17	216.0	433.1	649.7	866.2	1,082.8
	20	254.8	509.6	764.3	1,019.1	1,273.9
POWER,	23	293.0	586.0	879.0	1.172.0	1,465.0
WATTS	26	331,2	662.4	993.6	1,324.8	1,656.1
	30	382.2	764.3	1,146.5	1,528.7	1,910.8
	40	509.6	1,019.1	1,528.7	2,038.2	2,547.8
	50	637.0	1,273.9	1,910.8	2,547.8	3,184.7
	60	764.3	1,528.7	2,293.0	3,057.3	3,821.7
	70	891.7	1,783.4	2,675.2	3,566.9	4,458.6
	80	1,019.1	2,038.2	3,057.3	4,076.4	5,095.5
	90	1,146.5	2,293.0	3,439.5	4,586.0	5,732.5

Note: $\text{Spot Area} = \pi r^2 = \pi(0.05\ cm)^2 = 7.85 \times 10^{-3}\ cm^2$

$$\text{Energy Density} = \frac{(\text{power})(\text{pulse duration})}{\text{spot area}}$$

$$= \frac{(\text{Watt})(\text{second})}{cm^2} = \frac{\text{Joules}}{cm^2}$$

TABLE X (CONTINUED)

ENERGY DENSITIES, J/cm^2

		PULSE DURATION, SECONDS				
		0.6	0.7	0.8	0.9	1.0
	5	382.2	445.9	509.6	573.2	636.9
	8	611.5	713.4	815.3	917.2	1,019.1
	11	840.8	980.9	1,121.0	1,261.1	1,401.3
	14	1,070.1	1,248.4	1,426.8	1,605.1	1,783.4
	17	1,299.4	1,515.9	1,732.5	1,949.0	2,165.6
	20	1,528.7	1,783.4	2,038,2	2,293.0	2,547.8
POWER,	23	1,758.0	2,051.0	2,343.9	2,636.9	2,929.9
WATTS	26	1,987.3	2,318.5	2,649.7	2,980.9	3,312.1
	30	2,293.0	2,675.2	3,057.3	3,439.5	3,821.7
	40	3,057.3	3,566.9	4,076.4	4,586.0	5,095.6
	50	3,821.7	4,458.6	5,095.5	5,732.5	6,369.4
	60	4,586.0	5,350.3	6,114.6	6,879.0	7,643.3
	70	5,350.3	6,242.0	7,133.8	8,025.5	8,917.2
	80	6,114.6	7,133.8	8,152.9	9,172.0	10,191.1
	90	6,879.0	8,025.5	9,172.0	10,318.5	11,465.0

Note: Spot Area $= \pi r^2 = \pi(0.05\ cm)^2 = 7.85 \times 10^{-3}\ cm^2$

$$\text{Energy Density} = \frac{(\text{power})(\text{pulse duration})}{\text{spot area}}$$

$$= \frac{(\text{Watt})(\text{second})}{cm^2} = \frac{\text{Joules}}{cm^2}$$

FIGURES

Diagram of Specimen

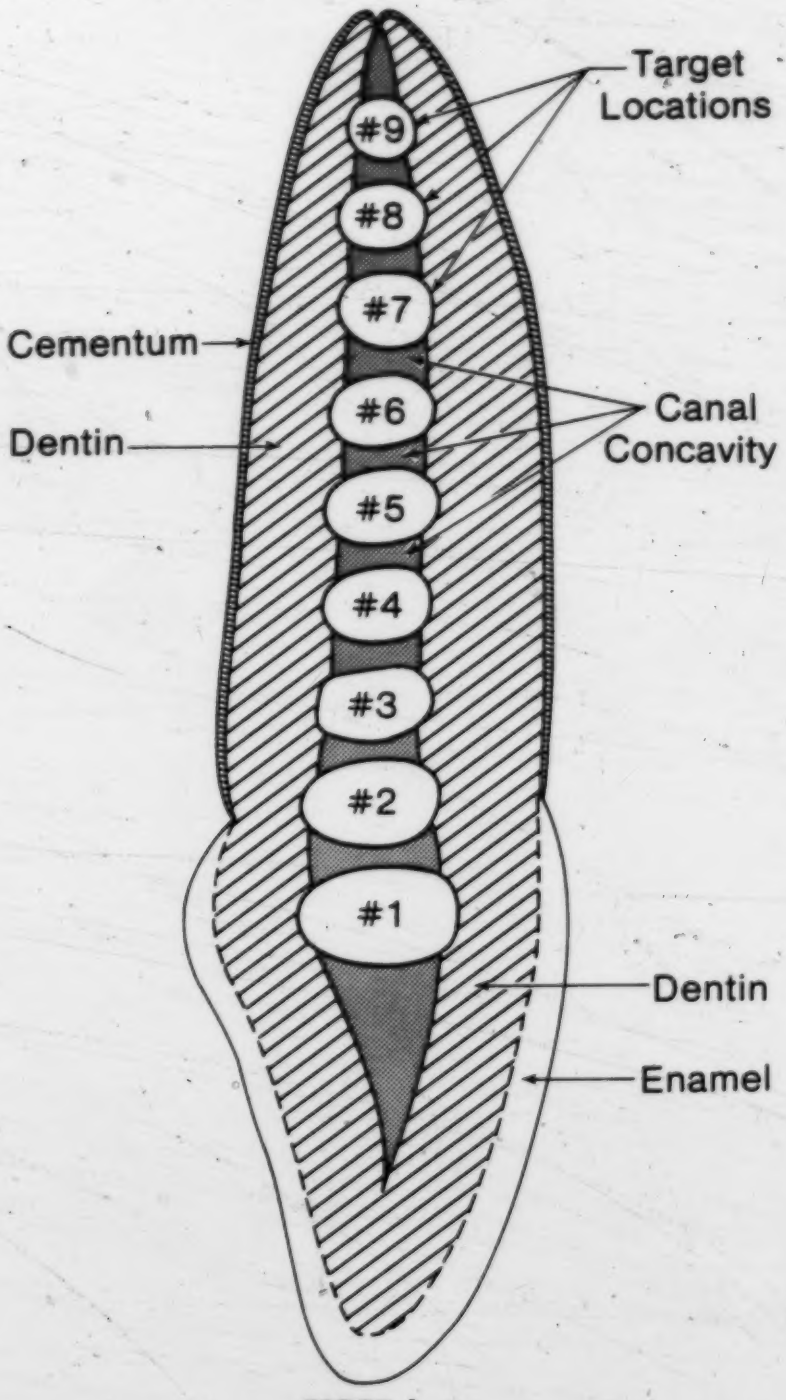


FIGURE 1

PHOTO NO.	FUSION COVERAGE		THERMAL CRACKING	
	NONE TO SLIGHT	MODERATE TO TOTAL	NONE TO SLIGHT	MODERATE TO SEVERE
1	____	____	____	____
2	____	____	____	____
3	____	____	____	____
4	____	____	____	____
5	____	____	____	____
6	____	____	____	____
7	____	____	____	____
8	____	____	____	____
9	____	____	____	____
10	____	____	____	____
.	.	.	.	.
.	.	.	.	.
.	.	.	.	.
.	.	.	.	.
.	.	.	.	.
.	.	.	.	.
98	____	____	____	____

FIGURE 2

Evaluation form used to categorize

laser-induced effects on dentin

POWER, WATTS

PULSE DURATION, SECONDS	1 to 10	11 to 15	16 to 20	21 to 26	27 to 50
0.1					
0.2					
0.3					
0.4					
0.5					
0.6					
0.7					
0.8					
0.9					
1.0					
TOTAL					

ATTRIBUTE TESTED:

AMOUNT OF FUSION COVERAGE

FIGURE 3

PULSE DURATION, SECONDS

POWER, WATTS	0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9	1.0
1 to 10										
11 to 15										
16 to 20										
21 to 26										
27 to 50										
TOTAL										

ATTRIBUTE TESTED:

AMOUNT OF FUSION COVERAGE

FIGURE 4

POWER, WATTS

PULSE DURATION, SECONDS	1 to 10	11 to 15	16 to 20	21 to 26	27 to 50
0.1					
0.2					
0.3					
0.4					
0.5					
0.6					
0.7					
0.8					
0.9					
1.0					
TOTAL					

ATTRIBUTE TESTED:

AMOUNT OF THERMAL CRACKING

FIGURE 5

PULSE DURATION, SECONDS

		0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9	1.0
POWER, WATTS	1 to 10										
	11 to 15										
	16 to 20										
	21 to 26										
	27 to 50										
	TOTAL										

ATTRIBUTE TESTED:

AMOUNT OF THERMAL CRACKING

FIGURE 6

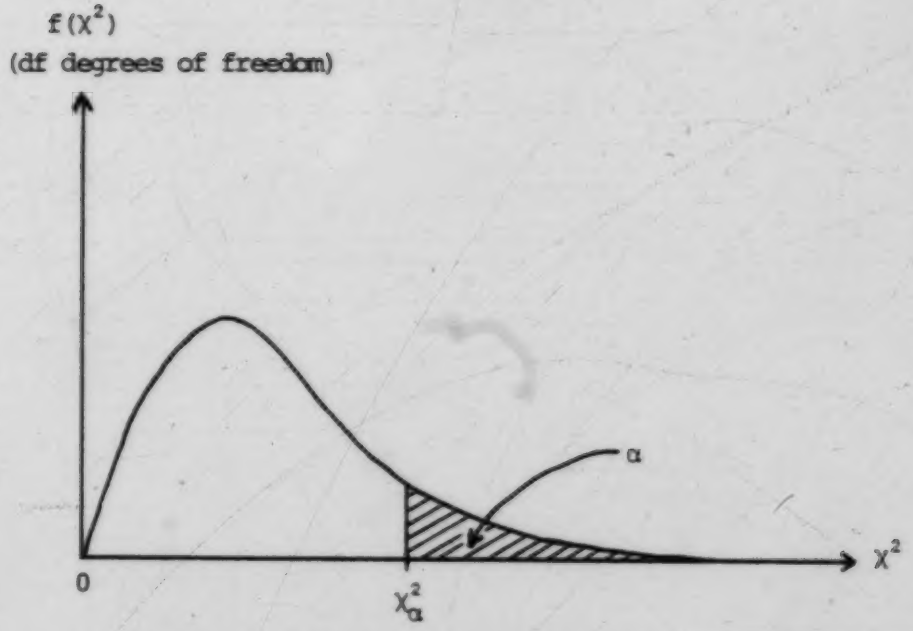
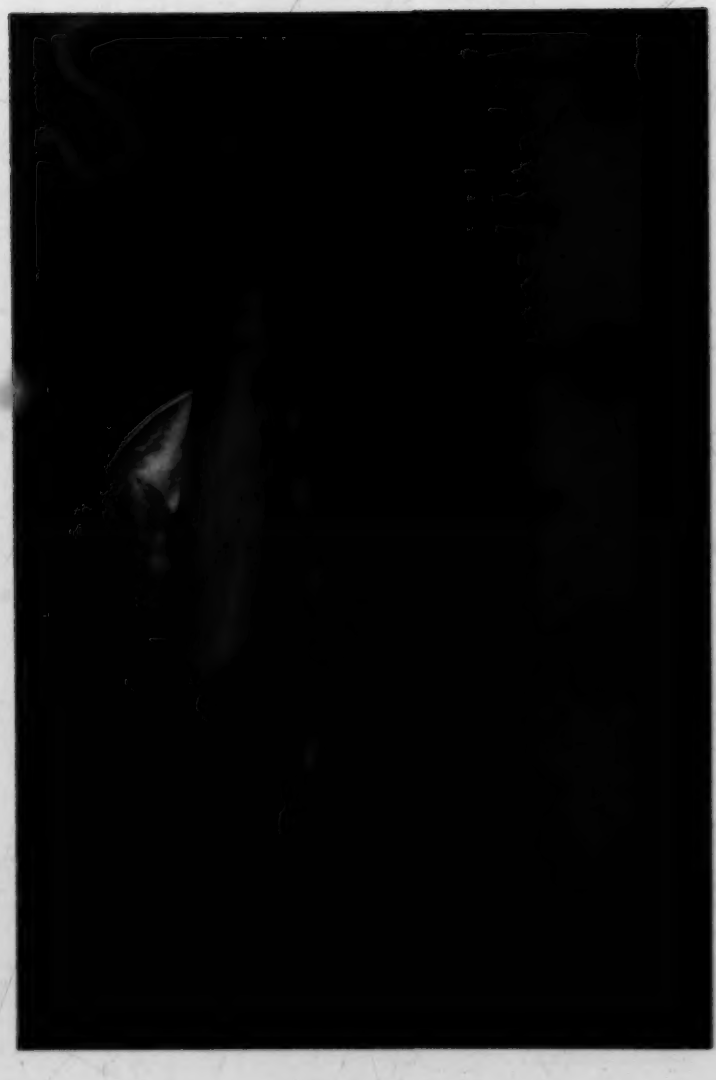


FIGURE 7

Chi-square distribution

PLATES

PLATE 1



Photograph of undyed split tooth

specimen showing target areas

COLOURED PICTURES
Images en couleur

PLATE 2

Photograph of distal tip of fiber held 2 mm

away from surface of target area

COLOURED PICTURES
Images en couleur

PLATE 3

Photograph of Nd-YAG laser used:

Molectron Model 5000 Photocoagulator

COLOURED PICTURES
Images en couleur

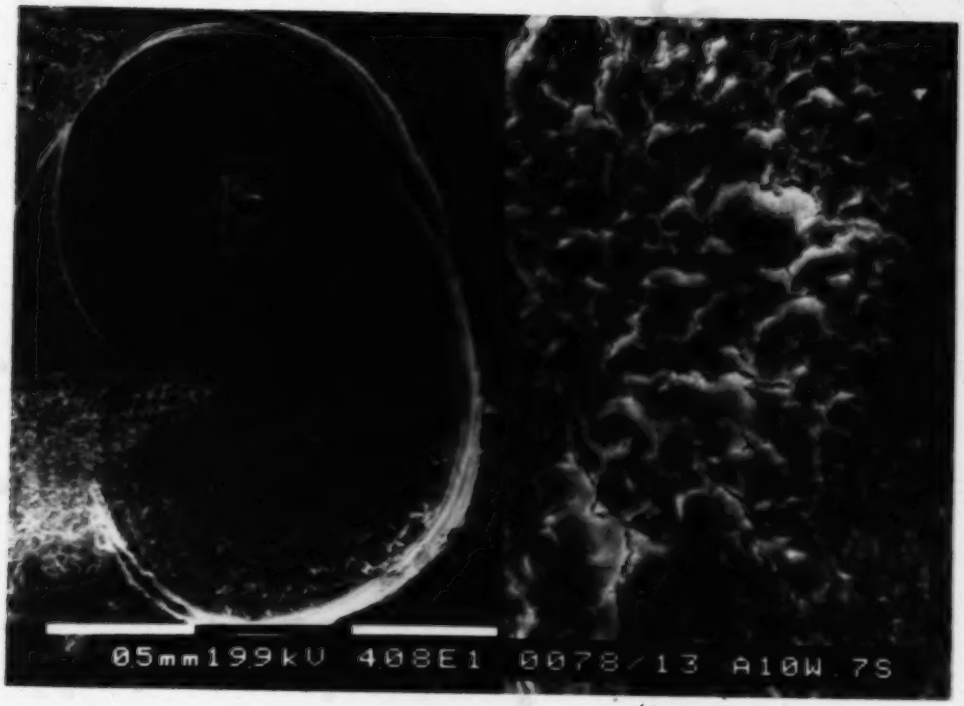
PLATE 4

Photograph of dyed split tooth

specimen showing target areas

COLOURED PICTURES
Images en couleur

PLATE 5



SEM of a laser induced disruption

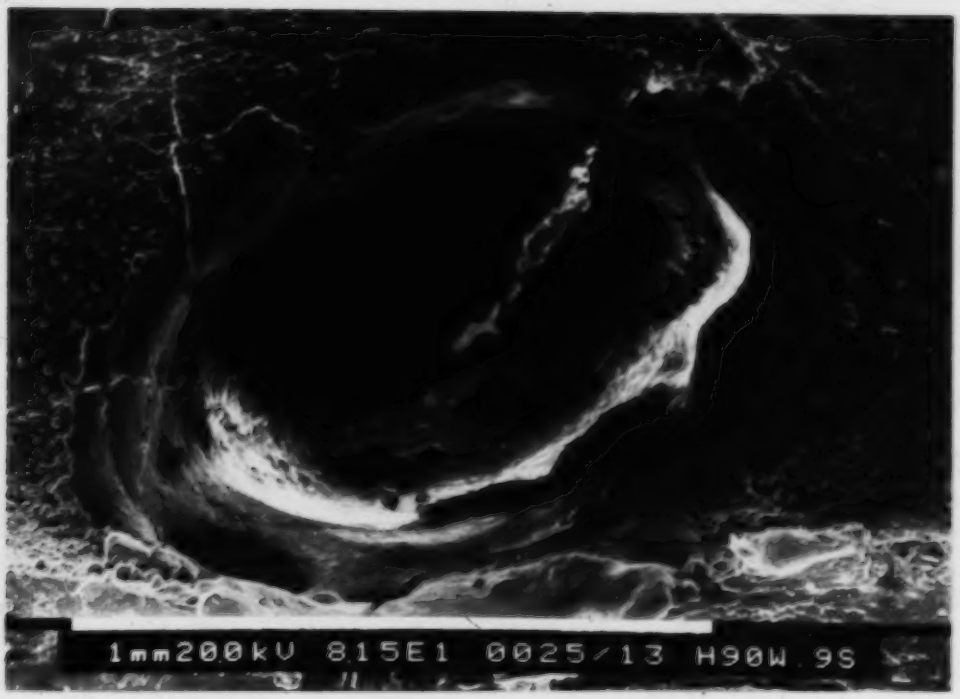
of the smear layer

PLATE 6

Color photograph of laser impact site in an undyed specimen showing dentinal fusion, recrystallization, and minimal thermal cracking

COLOURED PICTURES
Images en couleur

PLATE 7

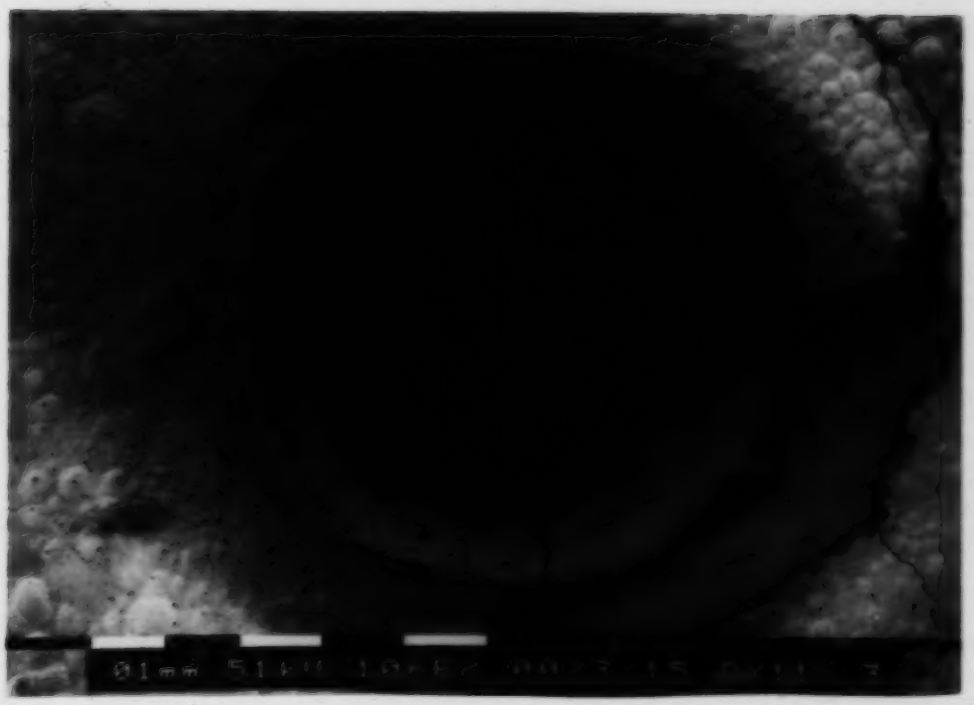


Scanning Electron Micrograph of laser impact area in undyed specimen

PLATE 8

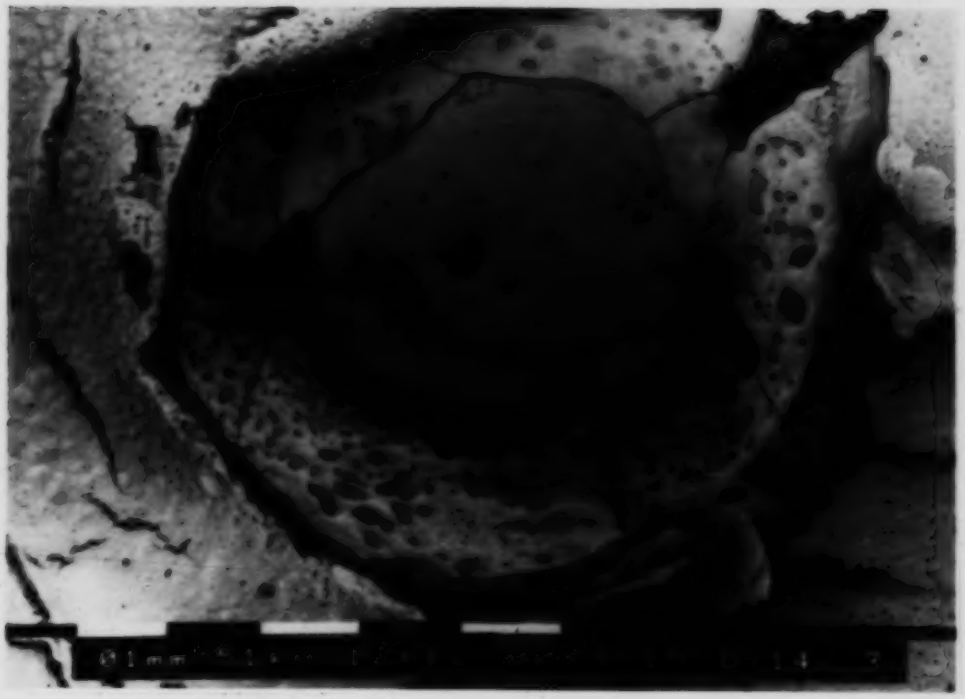
Scanning Electron Micrograph (2020X) of melted and recrystallized dentin in an undyed specimen

PLATE 9



Scanning Electron Micrograph of laser impact area in dyed specimen demonstrating minimal thermal cracking and moderate to total fusion coverage

PLATE 10



Scanning Electron Micrograph of laser impact area in dyed specimen demonstrating moderate to severe thermal cracking and moderate to total fusion coverage

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APPENDIX

APPENDIX

A prerequisite to understanding the Friedman's nonparametric analysis of variance is an understanding of the chi-square distribution. This is because statistical inferences from the results of the Friedman's test are allowed on the basis that its distribution approximates that of the chi-square distribution for large enough sample sizes.

The chi-square distribution describes the theoretical sampling distribution of s^2 (the variance of the sample) for random samples from normal populations. More specifically, we have:

THEOREM: If s^2 is the variance of a random sample of size n taken from a normal population having the actual population variance σ^2, then

$$\chi^2 = \frac{(n - 1)s^2}{\sigma^2}$$

is a value of a random variable having the chi-square distribution with the degrees of freedom df = n - 1.

In simpler terms, the chi-square test allows for the comparison of the sample variance actually obtained experimentally (s^2) with the expected value for the variance, the population variance (σ^2).

A table is widely available which contains selected values of χ^2_α for various values of df, the number of degrees of freedom, where χ^2_α is such that the area under the chi-square distribution to its right is equal to , the significance level. The general shape of the chi-square distribution is shown in Figure 7.

Usually the value obtained for the variance of the sample will differ from the actual variance of the population from which the sample was drawn. The chi-square test can help to determine what this means. This test holds the null hypothesis (H_0) to be that the sampling variance (s^2) is close enough to the expected value of σ^2 to be within a reasonable expected range if the sample was actually taken from the population of variance σ^2. In other words, H_0 states that the sample was likely taken from the population with variance σ^2. The alternative hypothesis (H_1) states that s^2 is too far away from σ^2 to likely have been taken from the population with variance σ^2. This implies that chance alone was not likely responsible for this difference and some other factor may be responsible.

This significance level, α, also represents the probability of committing a Type I error; that is, of rejecting H_0 when in fact it is true. Since rejecting H_0 by default means that you accept H_1, committing a Type I error means that you conclude that some factor other than chance had a significant influence on the data

when it actually did not. Because of the nature of this experimental study, it is desirable to minimize the chance of committing a Type I error. Accordingly, 0.05 was chosen to represent the significance level. This means that it was decided beforehand to require that the chance of incorrectly concluding that factors other than pure chance influenced the data be 5 out of 100 or less. In theory, then, for any given n samples with sample variance s^2 taken from a normal population with the variance of σ^2 a chi-square value can be calculated which will correspond to a given significance α_{sample}. This α_{sample} is then compared to that previously chosen to be the level required for significance, α. Only if α_{sample} is equal to or less than can it be said that the sample variance, s^2, is "significantly" different than the actual population variance, σ^2, i.e. that the sample was not likely to have been drawn from the population with variance σ^2.

Rationale

The data encountered in this study was categorical in nature. It was possible to separate the photographs into categories by the amount of thermal cracking and fusion coverage. Due to the nature of the data, more powerful measurements (such as the length, width and number of cracks) were not feasible. The categories chosen also demonstrate that they have a particular relation to one another. One category is labled "no to minimal thermal cracking" and one is "moderate to severe thermal cracking". Their relation is that they both describe the amount of thermal cracking. The

same is true for the categories describing fusion coverage. Data of this kind can be classified as "ordinal" data.

When the data from k matched samples are in an ordinal scale, the Friedman two-way analysis of variance by ranks is useful for testing the null hypothesis that the k samples have been drawn from the same population. The data in this study can be considered to be "matched" because all power groupings were irradiated ten times (from 0.1 to 1.0 seconds at 0.1 second intervals).

Method

For the Friedman test, the data are cast in a two-way table having N rows and k columns. In one configuration, the rows represent the pulse durations and the columns represent the power intervals (see Figures 3 and 5). In the other configuration, the rows represent the power intervals and the columns represent the pulse durations (see Figures 4 and 6). The table configuration shown in Figures 3 and 5 organizes the data such that the Friedman test can test whether the groups of samples having the same pulse duration (regardless of the power used) are drawn from the same population. The table configuration shown in Figures 4 and 6 allows the Friedman test to examine whether the groups of samples irradiated at power levels within the given range (regardless of the pulse duration) are drawn from the same population. The Friedman test was run once for each of these tables, resulting in a total of four runs. In essence, it was expected that these four

runs would indicate whether variations in power level or pulse duration had a significant effect on the amount of resulting thermal cracking or fusion coverage.

After being organized into these tables, the data are ranked from 1 to k within each row. Then the verticle sums for each of the columns are calculated. These verticle sums can collectively be referred to as R_j where $j = 1$ to k and k is the number of columns.

Now if the null hypothesis (i.e. all the samples -columns- came from the same population) is in fact true, then the distribution of ranks in each column would be a matter of chance. Thus it would be expected that the ranks of 1,2,3,...k would appear in all columns with about equal frequency. This would result in the rank totals for the various columns being about equal. If the amount of thermal cracking or fusion coverage were dependent on the power or pulse duration (i.e., if H_0, the null hypothesis, were false) then the rank totals would vary from one column to another.

The Friedman test determines whether the rank totals (R_j) differ significantly. To make this test, the value of a statistic χ^2_r is computed.

When the number of rows and/or columns is not too small, it can be shown[70] that χ^2_r is distributed approximately as chi-square with df = k - 1, when

$$\chi^2_r = \frac{12}{Nk(k+1)} \sum_{j=1}^{k} (R_j) - 3N(k+1)$$

where N = number of rows

k = number of columns

R_j = sum of ranks in jth column

Inasmuch as the sampling distribution of χ^2_r is approximated by the chi-square distribution with df = k - 1, the probability associated with the occurrence under H_0 of values as large as χ^2_r is shown in the previously mentioned table of chi-square (χ^2) values. If the value of χ^2_r as computed by the above formula is equal to or larger than that given in the table of chi-square values for a particular level of significance and a particular value of df = k - 1, the implication is that the sum of the ranks (or, equivalently, the mean ranks, R_j/N) for the various columns differ significantly (which is to say that the amount of thermal cracking and/or fusion coverage depends on the laser exposure parameters) and thus H_0 may be rejected at that level of significance.